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## Fibre optic force sensor for flexible bevel tip needles in minimally invasive surgeries

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**Abstract:** This paper presents and discusses the design of a novel fibre optic force sensor which will be embedded within the needle. This sensor will provide information on lateral and axial forces acting on the needle. The proposed design is amalgamation of two technologies and uses both Fabry Perot interferometer and fibre Bragg grating.

**Keywords:** flexible needles; minimally invasive; FBG sensors; FPI sensors; hybrid sensors; force sensing; fibre optic.

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### 1 Introduction

*Minimally invasive surgeries (MIS)* using flexible bevel tip needles is active area of research, the applications for these include Brain, liver and prostate surgery, blood/fluid sampling, regional anaesthesia and catheter insertion to name a few applications (Goksel et al., 2005; Fichtinger et al., 2007; Wei et al., 2004a, 2004b; Fichtinger et al., 2006;

Wan et al., 2005; Kettenbach et al., 2005a, 2005b; Meltsner et al., 2006; Kennedy et al., 2006; Yu et al., 2007). Accurately planning and executing the optimal motions for such steerable needles is difficult and has some associated complications:

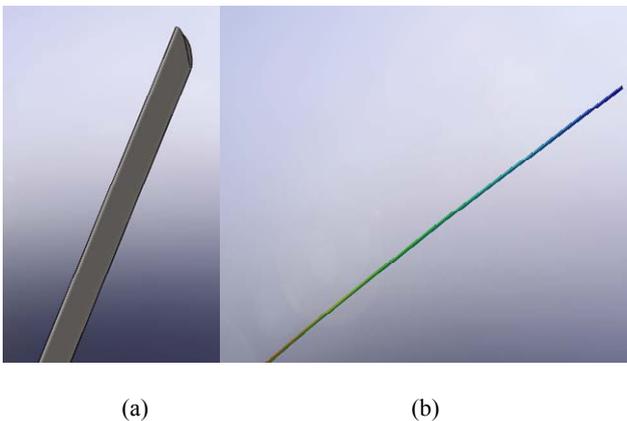
- torsion
- tissue in-homogeneity, i.e., forces acting on the needle

- image feedback and planning of bevel tip position to help plan the steering.

The topic of force sensing is a category of scientific and engineering problem, in our case the goal is to obtain data from the physical operation of the needle inside the tissue while surgery. In most of the minimally invasive procedures the force applied to an organic tissue can only be estimated through visual feedback by observing the deformation of the tissue. Such lack of the sensing ability greatly limits effectiveness of the operations and is considered a major downside of the current robotic systems. Also the obstructions (tumours, nerves, muscles, etc.) buried under the tissue surface may not be discovered even with advanced visual devices, but it could be detected by a simple sense of touch. There is a critical need for methods to quickly, accurately and easily manipulate instruments within closed visual feedback. Catheter, probe and needle interventions can benefit from system which measure tool position and forces in real-time.

To successfully understand tissue needle interaction many Finite element models (FEM) have been proposed (Azar et al., 2000; Misra et al., 2008; Abolhassani et al., 2007) which have provided a good understanding about the tissue needle interactions. It has been established that the force distribution along the needle is not uniform. A small experimentation when a needle is put under stress produces the results as shown in Figure 1(b), it is evident from the results that the stresses on needles are not uniformly distributed.

**Figure 1** (a) Bevel tip needle (b) Force distribution on the needle when under stress (see online version for colours)



The force sensors designed for surgical instruments need to have special attention paid to them as the size of the sensor must not exceed the size of the standard instrument insertion through which it is to be inserted during surgery. The sensor should be completely sealed to prevent body liquids affecting internal sensing elements. The sensor should be sterilisable with a standard sterilisation protocol. Also in this case it is very desirable for the sensor to be magnetic resonance imaging (MRI) compatible.

Fibre optics have distinct advantages over traditional sensors as they are intrinsically safe, passive: no need for

electrical power at sensor end, possibility of remote, multiplexed operation, small size and lightweight, integrated telemetry: fibre itself is data link, wide bandwidth, high sensitivity. Many fibre optic-based systems have been used for catheters (Polygerinos et al., 2009; Puangmali et al., 2008). Following are the expectations from the sensor:

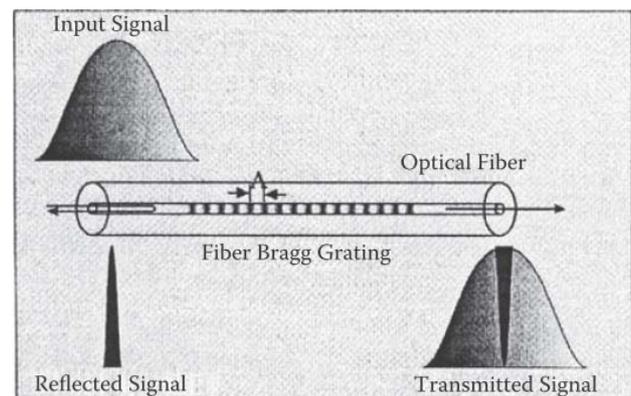
- it will provide real-time force measurement of flexible bevel tip needles
- bending/position information
- frictional forces along the body of the needle.

## 2 Fibre optic sensor

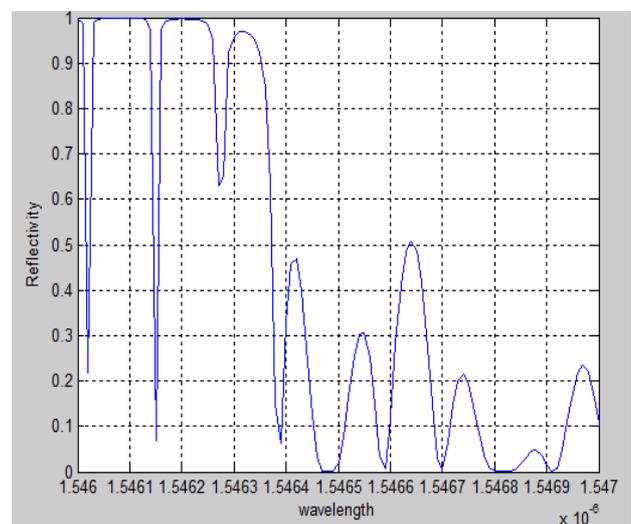
### 2.1 Fibre bragg grating

Fibre bragg grating (FBG) has been used in many applications to detect strain and temperature (Rajan et al., 2010; Yokoyama et al., 2008; James et al., 1996; Yi et al., 2007; Wang, 2005; Li et al., 2010). The sensor has been used for finding lateral forces in many medical instruments, typically in catheters. As shown in Figure 2 typical strain response of a FBG sensor.

**Figure 2** (a) FBG sensor (b) Reflection response from a FBG sensor under no strain (see online version for colours)



(a)



(b)

Our design involves use of tilted fibre brags grating (TFBG), this grating unlike the normal FBG, propagate in both core and cladding and provide information on direction along with the bend. In addition to this TFBG have an inherent property of temperature compensation (Li et al., 2010).

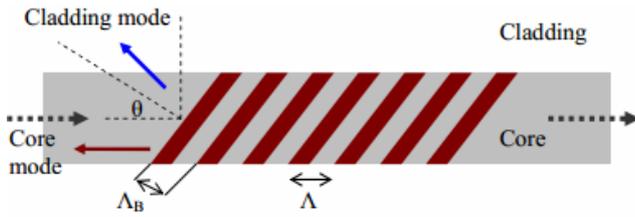
### 2.1.1 Working principle

The working principle of TFBG is based around reflective signal from core and cladding, reflected signal changes based on the both magnitude and direction of bending.

$$\lambda_i = \left( {}^{co}n_{eff} + {}^{clad}n_i \right) \cdot \frac{\lambda}{\cos\theta} \quad (1)$$

where  $\lambda_i$ ,  ${}^{co}n_{eff}$ ,  ${}^{clad}n_i$ ,  $\theta$ ,  $\lambda$  are cladding mode resonance wavelength, effective index of the core, effective index of the cladding mode, blaze angle and grating period respectively for a TFBG as shown in Figure 3.

**Figure 3** TFBG sensor and concerning parameters (see online version for colours)



When strain is applied the reflected wavelength shifts, which is defined by the following expression:

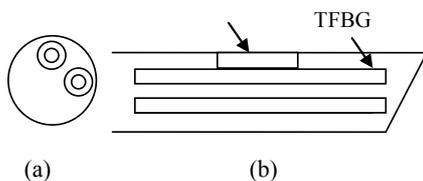
$$\begin{pmatrix} \Delta\lambda_{core} \\ \Delta\lambda_{cladding} \end{pmatrix} = \begin{pmatrix} P_{Core,\varepsilon} & P_{Core,T} \\ P_{Clad,\varepsilon} & P_{Clad,T} \end{pmatrix} \begin{pmatrix} \varepsilon \\ T \end{pmatrix} \quad (2)$$

where  $\Delta\lambda$  the shift is in the reflected wavelength,  $P_{Clad,\varepsilon}$   $P_{Clad,T}$  represents the strain and temperature response of the core,  $P_{Core,\varepsilon}$   $P_{Core,T}$  represent the strain and temperature response of the cladding.

### 2.1.2 Practical implementation

The design of the sensor is inspired from Su et al. (2011a, 2011b), the design involves use of two TFBG which are 90 degrees apart (which are to be placed in the needle); this part of the design is purely for finding the 3-D profile of the flexible needle while it is in the tissue.

**Figure 4** (a) Showing the top view of the sensor (b) Showing the front view



In the above case when the needle bends the TFBG the reflected signal from core and cladding changes, which is measured by a FBG interferometer, the change in the wavelength from core directly corresponds to the strain and the wavelength response (in addition to the propagating modes) enable us to find the direction of the bend.

It is envisaged that the sensors in the needle will be able to provide us with bending information (stress) along with the forces acting on the tip, this will not only help in determining the overall forces acting on the needle but also help in finding frictional forces (when combined with the second sensor as described in the next section).

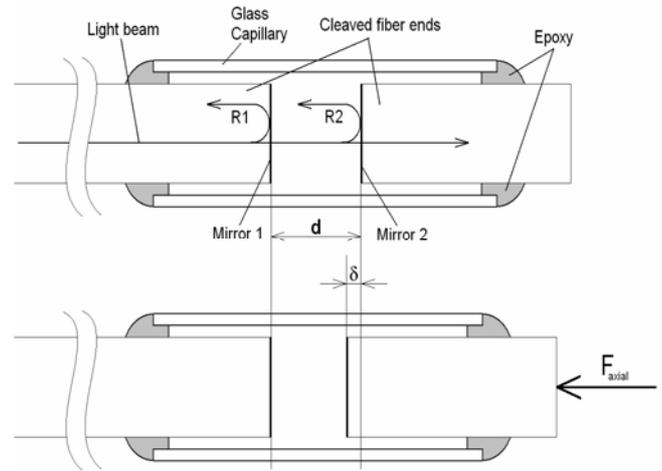
## 2.2 Fabry-Pérot interferometry

Fabry-Pérot interferometry-based (FPI) fibre optic sensors have been used in many applications before either as strain, force, temperature, displacement or refractive index sensor. Su et al. (2011a, 2011b) describe the use of an FPI sensor to measure forces acting on a needle tip in brachytherapy. The appeal of this technique is its small size (a single fibre with a small sensing cavity at one fibre end) and relatively simple fabrication process.

### 2.2.1 Working principle

The working principle of the Fabry-Pérot interferometer is based on a change in optical path length in back-reflecting light beams in a single fibre. The small deformable cavity at the end of the fibre (see Figure 5) consists of two (partially translucent) mirrors. Mirror 1 is the reference mirror and will reflect part (R1) of the light beam coming from the fibre. The transmitted part of the light beam falls onto mirror 2 and again, part of the light (R2) will be reflected and a part will be transmitted. The two reflected light beams each have a different optical path length thus a different phase. This will create interference fringes.

**Figure 5** Schematic representation of the FPI sensing cavity



Note: The top shows the unloaded case, the bottom shows a situation where a load is applied.

When an axial force is applied to the tip of the sensor, the distance between the two mirrors will change a distance  $\delta$ . Therefore, the phase shift of R2 relative to R1 will change. The intensity of the combined light rays returning can be measured and used as an indication for the force.

Phase shift  $\Delta\phi$  and intensity  $I$  are related in the following sense:

$$I = I_1 + I_2 + 2 \cdot \sqrt{I_1 \cdot I_2} \cdot \cos(\Delta\phi)$$

With  $I_i$  the intensity of beam  $i$ . The phase shift can be expressed in terms of the geometry in the following way.

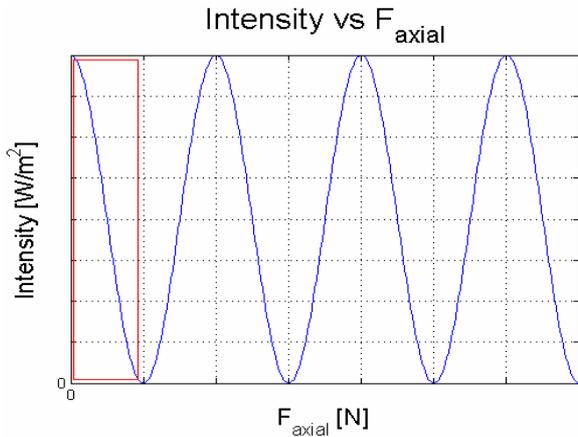
$$\Delta\phi = \frac{2\pi \cdot \delta}{\lambda} = \frac{2\pi \cdot \varepsilon_x \cdot d}{\lambda}$$

With  $\varepsilon_x$  the strain in axial direction,  $d$  the distance between the two mirrors and  $\lambda$  the wavelength of the light source used. Combining these two equations gives

$$I = I_1 + I_2 + 2 \cdot \sqrt{I_1 \cdot I_2} \cdot \cos\left(\frac{2\pi \cdot \varepsilon_x \cdot d}{\lambda}\right)$$

An applied force will induce a strain of the sensing cavity and this in turn will induce a change in intensity that can be measured using a photo diode. A typical intensity curve resulting from the interference would look like Figure 6, with maxima and minima occurring periodically.

**Figure 6** Example of an intensity graph to illustrate the relation between force and intensity using an FPI sensor (see online version for colours)



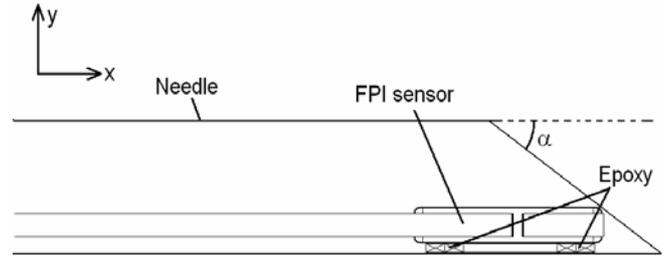
For ease of use in the measurements it is convenient to have 1 intensity value corresponding to 1 force value without the need to count maxima and minima, hence to only make use of the area indicated with the red rectangle. This implies the maximum deflection of mirror 2 to be  $\delta = \frac{1}{4}\lambda = 387.5$  [nm] under the maximal load occurring during operation.

### 2.2.2 Practical implementation

Forces acting on the needle tip need to be measured as closely to the tip as possible to ensure the desired force is

measured. To accomplish this, the FPI sensor can be attached to the needle using epoxy as shown in Figure 7.

**Figure 7** The implementation of the FPI sensor in the tip of the needle

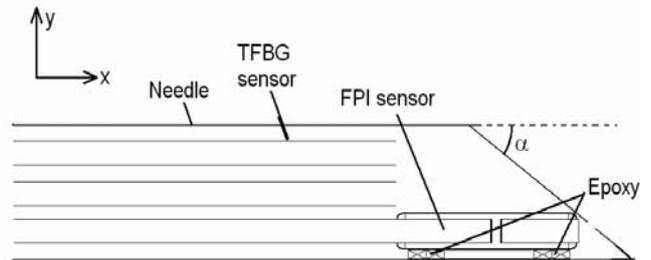


When the distance between the mirror and light increases the intensity of light being received by the photodiode increases, on performing the fast Fourier transforms (FFT), the signals can be resolved and adequate measurement on strain can be performed.

## 3 Design

The application of sensing forces on the needle gratings are to be placed at three different positions and are going to be three different wavelengths, the grating has to be long. This will help provide 3-D bend information of the sensor as described in James et al. (1996). In addition to this when depending on load on the tip of the needle further signal will be reflected back by FPI sensor. The proposed design is as shown in Figure 8.

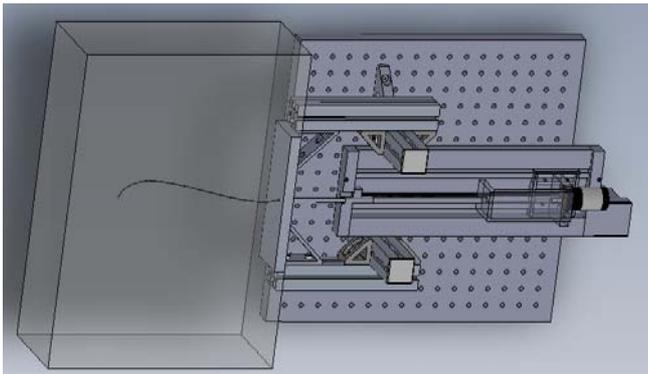
**Figure 8** Sensor design, which is to be placed within the needle



## 4 Testing system

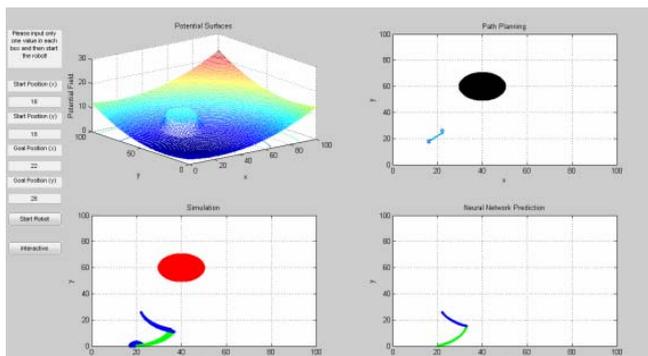
For testing of the system, a robot has been designed in Figure 9, which uses a linear actuator and a DC motor to translate and rotate the needle into phantom tissue, the needle is a hollow nitinol wire (it is hollow to allow the fibre to be fitted inside it). The diameter of the needle is 0.3 mm with a bevel angle of 45°.

**Figure 9** The test rig for testing the sensor inside phantom tissue (see online version for colours)



As the needle is steered inside the phantom tissue two cameras will validate the forces which will act on the needle. The control of the robot is being done using a data acquisition card (USB-128), the control software for this is being written in MATLAB. The path planning for this done using algorithms based on Dubins path. The path planning is done using MATLAB – shown on Figure 10.

**Figure 10** The simulations of the needle control algorithm (see online version for colours)



It has been established that the overall forces ( $F_T$ ) acting on the needle can be described as a summation of frictional force ( $F_f$ ), stiffness force ( $F_s$ ) and cutting force ( $F_c$ ).

$$F_{Total} = F_f + F_c + F_s$$

In the above equation cutting force is a constant, where as the stiffness force and frictional force acting on the needle are variables and depend on the tissue. Using the proposed sensor in the above sections we can find the stiffness force acting on the needle, this will not only help in estimating complex structures of tissue but will also help in calculations of frictional forces acting along the body of the needle.

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